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DOI:

[10.1177/0954411913483260](https://doi.org/10.1177/0954411913483260)

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Document Version

Peer reviewed version

Citation for published version (Harvard):

Parekh, J, Shepherd, DET, Hukins, DWL, Hingley, C & Maffulli, N 2013, 'In vitro investigation of friction at the interface between bone and a surgical instrument', *Institution of Mechanical Engineers. Proceedings. Part H: Journal of Engineering in Medicine*, vol. 227, no. 6, pp. 712-8. <https://doi.org/10.1177/0954411913483260>

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In vitro investigation of friction at the interface between bone and a surgical instrument

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Author generated final version. Published as:

Parekh J, Shepherd DET, Hukins DWL Hingley C & Maffulli N (2013). In vitro investigation of friction at the interface between bone and a surgical instrument. *Journal of Engineering in Medicine* 227: 712–718.

Abstract

This study investigated the friction between surgical instruments and bone to aid improvements to instrument design. The bases of orthopaedic surgical instruments are usually made of metal, especially stainless steel. Silicone elastomer was chosen as an alternate biocompatible material, which would be compliant on the bone surface when used as the base of an instrument. The coefficient of static friction was calculated at the bone/material interface in the presence of a synthetic solution that had a comparable viscosity to that of blood, to assess the friction provided by each base material. Three types of silicone elastomer with different hardnesses (Shore A hardness 23, 50, and 77), and three distinct stainless steel surfaces (obtained by spark erosion, sand blasting and surface grinding) were used to assess the friction provided by the materials on slippery bone. The bone specimens were taken from the flattest region of the femoral shaft of a bovine femur; the outer surfaces of the specimens were kept intact. In general, the stainless steel surfaces exhibited higher values of coefficient of static friction, compared to the silicone elastomer samples. The stainless steel surface finished by spark erosion (surface roughness $R_a = 8.9 \pm 1.6 \mu\text{m}$) had the highest coefficient value of 0.74 ± 0.04 . The coefficient values for the silicone elastomer sample with the highest hardness (Dow Corning Silastic Q7-4780, Shore A hardness 77) was not significantly different to values provided by the stainless steel surface finished by sand blasting (surface roughness $R_a = 2.2 \pm 0.1 \mu\text{m}$) or surface grinding (surface roughness $R_a = 0.1 \pm 0.0 \mu\text{m}$). Based on the results of this study it is concluded that silicone could be a potentially useful material for the design of bases of orthopaedic instruments that interface with bone.

Key words: Bone; friction coefficient; silicone; stainless steel; surgical instrument

Introduction

The bases of some orthopaedic surgical instruments interface with the curved surfaces of bone. Examples of such instruments include the angle guide for the treatment of fractures of the proximal femur with a dynamic hip screw.¹ The angle guide is temporarily attached to the upper end of the femur, to accurately provide both horizontal and vertical reference required by the surgeon to insert the guide pin into the femoral head. Typically these instruments are manufactured from stainless steel.

A number of studies have investigated friction at the interface between implants and bone. Rancourt *et al.*² investigated the friction properties of the interface between porous-surfaced metals and tibial cancellous bone and found porous surfaces to have a higher coefficient of friction. Friction coefficients varied between 0.3 and 1.3. Dammak *et al.*³ investigated the frictional forces between different metal surfaces and polyurethane or bone and showed that a textured metal surface had the highest coefficient of friction. Davim and Marques⁴ investigated the friction between bovine cancellous bone and steel with a lubricant of water. An average friction coefficient of 0.25 ± 0.02 was found. The surfaces used during these tests were dry or water lubricated. In practice, a bone surface is likely to be covered in blood, leading to a reduced coefficient of friction when contacted by a surgical instrument. These studies show that surface texture can affect friction between metals and bone and that there is a lack of information about friction between materials and bone when the surface is covered with blood.

While these studies on the interface between implants and bone have been undertaken, there are no studies that have specifically investigated the friction between a surgical instrument and bone.

An alternate approach to using stainless steel for the bases of surgical instruments would be to use a compliant material that conforms to the bone surface when the surgeon pushes on

the instruments and potentially increases the friction between the instrument and the bone. This approach has been used for bone plates which have been modified by inserting silicone elastomer sheets in between the plate and the bone.⁵ As a result, *in vitro* plate-bone contact was greater and interface pressure was lower as compared with a standard plate. In this study silicone elastomers were chosen as an example of compliant material as they are biocompatible materials and easily processed.⁶ Silicones can withstand sterilisation with irradiation (for a single use instrument) or by autoclaving (for a multiple use instrument).⁷

The purpose of this study was to determine whether textured stainless steel produced by conventional metal finishing processes or a silicone elastomer would provide a higher coefficient of static friction at the bone surface, leading to a increased friction by the device on the bone. Standard textured stainless steel surfaces that had been roughened by surface grinding, sand blasting, and spark erosion were compared with three different grades of silicone elastomers, of varying Shore A hardness.

Materials and Methods

Bone Specimen

Three cortical bone specimens (approximate dimensions of 25 mm × 15 mm × 5 mm) were cut from the flattest region of the femoral shaft of a frozen bovine femur obtained from Fresh Tissue Supplies (Heathfield, East Sussex, UK). This region was at the lateral-proximal end of the bone. The bone was initially cut with a saw to achieve a roughly rectangular-shaped block. Subsequently a file was used on the inner sides of the cortical bone to obtain a specimen with the desired dimensions. The outer surface of the bone, to be used for the friction tests, was not cut or filed. Each sample was secured in an open steel cup (internal dimensions 31 mm × 26 mm × 5 mm) using acrylic cement (WHW Plastics, Hull, UK). The bone surface was levelled using a spirit level and protruded above the top of the cup by about 3 mm.

Stainless Steel Samples

Standard samples of finishes on stainless steel surfaces obtained by surface grinding, sand blasting, and spark erosion (surface dimension 30 mm × 30 mm, manufactured by Rubert & Co. Ltd. Cheadle, Cheshire, England) were used. Surface roughness (R_a) values were measured using a contact method (Form Talysurf-120L, Taylor-Hobson Ltd., Leicester, UK; calibrated to 0.14 μm) using a diamond tipped stylus (radius 2 μm) over a surface area of 1 mm^2 . The readings were taken 6 times on different regions of the surface to ensure that whole the surface was considered.

Silicone Elastomer Samples

Three different grades of silicone were used: Silastic Q7-4720, Silastic Q7-4750 and Silastic Q7-4780 (all from Dow Corning Ltd, Coventry, UK). All three were biomedical grades intended for fabricating medical devices, including those intended for implantation in humans for less than 30 days (manufacturer's data sheet). According to the data sheet, the softest silicone was Q7-4720 (Shore A hardness 23), followed by Q7-4750 (Shore A hardness 50) and Q7-4780 (Shore A hardness 77). The materials were supplied in two parts that were mixed in a Schwabenthan Berlin two-roll mill (Engelmann & Buckham Ltd, Alton, UK). Sheets of silicone (2 mm thick) were prepared in a Moor E1127 hot press (George E Moore & Sons Ltd, Birmingham, UK) under a 50 kN load (applied to an area of 175 mm × 150 mm) at a temperature of 116°C for 12 minutes, as recommended by the supplier. Further details on these materials and their preparation are given by Mahomed *et al.*⁸ Circular samples of 70 mm diameter were cut from these sheets using a template. Surface roughness values were measured for each grade of silicone sheet, using the same method, as described for the stainless steel samples.

Blood Analogue Solution

A blood analogue solution that mimics the viscosity of natural blood was used.⁹ The solution has been used in previous tests of medical devices.^{10,11} The solution was made by dissolving

xanthan gum (0.4 g; CAS – 11138-66-2, Sigma Aldrich, Town, Dorset, UK) and sodium chloride (5 g; table salt, J Sainsbury, London, UK) in glycerol (400 g; Fisher Scientific, Loughborough, Leicestershire, UK) and distilled water (594.6 g).⁹

Measurement of coefficient of friction

Measurement of the coefficient of friction was guided by an ASTM standard.¹² The equipment used is shown in Figure 1. The force to overcome the friction between the bone and the silicone or textured stainless steel was provided by a Bose ELF 3200 materials testing machine, operated under the control of WinTest software (Bose Corporation, ElectroForce Systems Group, Minnesota, USA). The machine was equipped with a load cell capable of measuring up to 225 N and a displacement transducer capable of measuring up to 13 mm. The force required to overcome friction and move the assembly (assembly of cylinder, open steel cup and bone specimen as shown in Figure 2) was applied by a nylon monofilament fishing line (Sunset line and twine, Kansas, USA; diameter 0.50 mm and capable of withstanding approximately 133 N). As the purpose of the study was to calculate the coefficient of static friction, a compliant force measuring system was very important.¹² Given its compliance, there was appreciable strain of the nylon monofilament before there was appreciable movement of the bone cup. This allowed the load cell to record the gradual increase in force, and finally the “breakaway force”, F' , at which the cup was able to overcome static friction and begin to slide. F' , was used to calculate the coefficient of static friction.

The nylon monofilament line was passed under a pulley between the cup and the actuator of the testing machine. The pulley was made of nylon (60 mm diameter, thickness of 9 mm, 0.50 mm wide with a 0.25 mm deep groove to accommodate the line) with a single row radial ball bearing (inner diameter 10 mm, outer race diameter 26 mm, sourced from RS Stock # 286-7568, RS Components Ltd., Northamptonshire, UK) in the centre to provide negligible friction. On the other side of the cup, the line passed over a second pulley and was tied to a

counter weight of 7.03 N (a weight of 5.01 N on a holder of weight 2.02 N). There were two reasons for having the counter weight: (1) to balance the upward force from the first pulley acting towards the cross-head of the machine; (2) to ensure that the assembly stood upright by providing tension in the line. Without the counter weight, the assembly may tip or fall during testing, because of the small dimensions (previously mentioned) of the bone at the bottom. Note that the load cell on the testing machine was set to zero after the counter weight was added. The height of the bench on which the assembly rested was adjusted to ensure that the lines were horizontal.

The resistance provided by the pulleys was negligible when compared to the force measured. To verify the presence of negligible resistance, a weight was hung from the pulley systems, using the same nylon line, and the force measured by the transducer was recorded. For weights of 20.05 N and 5.01 N (measured using an Ohaus GA200D balance, Ohaus, New Jersey, USA), the transducer recorded values of 20.01 N and 5.00 N, respectively.

Figure 2 shows an exploded view of the setup of the assembly. The open steel cup holding the bone specimen was screwed to a cylinder to form the assembly (assembly weighed 0.47 kg and masses were added in the range of 0.1 kg to 0.4 kg). This assembly was placed on top of the material (stainless steel or silicone elastomer) sample. The silicone elastomer was screwed on top of specimen bench whose height was adjustable. The stainless steel samples were attached to the bench using double-sided adhesive-tape. The absence of any undesirable movement (i.e., other than translation in the intended direction) was verified by manually pushing the assembly.

Before measuring the coefficient of friction, the bone surface was coated with the blood analogue solution, to mimic the lubrication of blood. A pipette was used to pour the solution (5 mL) on to the surface to be tested; a paintbrush (diameter 10 mm) was then used to evenly distribute the solution. Measurements were made at room temperature (20°C). The

testing machine was operated in displacement control (0.1 mm/s for a total displacement of 7 mm) and the force and displacement data were recorded at 200 Hz. The procedure measured the horizontal load-displacement response at the interface in the presence of a constant compressive force, W . Five different weights were used to achieve five different increasing values of W and each measurement was repeated five times, for a total of 25 measurements for each material combination.

Data Analysis

A plot of force against displacement was generated for each test run. The graphs were plotted using SigmaPlot 11 (Systat Software Inc., Chicago, Illinois, USA). F' defined as the force required for overcoming static friction and initiating slide, was considered to be the first maximum force in the plot. F' could be lower, higher, or even equal to the force needed to maintain the surface sliding in the subsequent relative motion.¹³ If static friction is much larger than kinetic friction, it is possible for a system to exhibit stick-slip.¹² The interface was considered to be stick-slip when the assembly repeatedly stuck to the interfacing material, followed by a slide of certain distance (slip) because of accumulation of force. During such an occurrence, F' was determined as the highest force recorded before the bone slipped on the material. F' was then plotted against the corresponding values of W and a regression line was fitted to the data for every bone-material pair. The value of the slope of the regression line was the coefficient of static friction, μ , for that interface.

Statistical calculations were performed using MINITAB Release 16 Statistical Software (Minitab Inc., Pennsylvania, USA). Normality of the distributions was assessed using the Anderson–Darling test.¹⁴ A one way analysis of variance (ANOVA) was undertaken using the Tukey method for multiple comparisons to investigate significant differences between the materials combinations. Results were considered significant for $p < 0.05$.

Results

Table 1 gives the surface roughness of the silicone elastomers and stainless steel surfaces used in these experiments. A typical plot of force against displacement for a silicone surface in contact with bone is given in Figure 3a. This interface, like all other silicone elastomer/bone pairs, did not exhibit a high F' compared to the force required to sustain slide. Figure 3b shows a typical plot of force against displacement for a textured stainless steel surface in contact with bone. This interface, like all other stainless steel/bone pairs, exhibited stick-slip behaviour, as described in the data analysis section.

The measured values of F' were then plotted against W , as shown in Figures 4a and 4b. In all cases there was a significant linear relationship between F' and W , with values of the squared linear correlation coefficient (R^2) ranged from 0.91 to 0.99 and there was no systematic displacement of data points from the line which was constrained to pass through the origin. The slope of this line is the coefficient of static friction, μ . Table 2 shows values of μ for each pair of bone/material combination.

For the silicone elastomers the Silastic Q7-4780 had a significantly greater coefficient of static friction ($p < 0.05$) compared with Silastic Q7-4720 and Silastic Q7-4750. For the stainless steel specimens, the only significant difference was that the coefficient of static friction for the spark eroded surface was significantly greater ($p < 0.05$) than that which had been textured by grinding. Comparison of the silicone elastomer and stainless steel specimens showed that the coefficient of static friction for the stainless steel surfaces was always significantly greater ($p < 0.05$) than for the elastomer specimens, with the single exception that there was no significant difference between results for Silastic Q7-4780 and a steel surface textured by grinding.

Discussion

The results of this study showed that textured stainless steel surfaces finished by sand blasting or spark erosion provide a higher coefficient of static friction against bone that is

coated with a blood analogue solution, than do silicone surfaces that conform to the bone. In this study the surfaces that were textured by grinding usually provided higher friction than silicone surfaces, with the exception of Silastic Q7 – 4780 which was the hardest of the silicones considered and, therefore, the least compliant. A stainless steel surface textured by spark erosion (surface roughness $R_a = 8.9 \pm 1.6 \mu\text{m}$) provided the highest friction. The corresponding value of the coefficient of static friction of 0.74 ± 0.04 was within the range of values reported for metal/bone interfaces in the absence of lubricant (range of 0.68 ± 0.10 to 0.94 ± 0.12).³

The small bone specimen size, restricted for the reasons given in the Materials and Methods section, limited the maximum compressive force that could be applied by a static weight to 8.55 N. The assembly would not stay upright if a larger weight was applied. However, the results from the experiment exhibited a linear relationship between W and F' . Surgeons can typically push down on an instrument by hand with a force of about 27 N.¹⁵ If we assume that the linear relationship between W and F' can be extrapolated (i.e., the interface continues to obey the laws of friction), the results of this investigation will be valid for interfaces between bone and surgical instruments.

Silastic Q7 – 4780 provided a high coefficient of static friction, which was not significantly different to that of the stainless steel surface that had been textured by grinding. However, the combination of a high coefficient of friction and appreciable compliance could have advantages as it would give the surgeon the flexibility to adjust an instrument on the bone without reduced friction of the instrument on a bone covered with blood. Bone plates have been modified by inserting silicone elastomer sheets in between the plate and the bone; as a result, *in vitro* plate-bone contact was greater and interface pressure was lower as compared with a standard plate.⁵ Bone plates with elastomer sheets could be particularly useful as bone plates are used to fix various fractures, particularly in weak (e.g., osteopenic) bones.

Selecting a material that had a greater compliance would reduce the stresses on the screws used to secure the plate to the bone.

No previous studies have specifically investigated the friction between a surgical instrument and bone, but the results of the current study can be compared with a number of studies that have investigated friction at the interface between implants and bone. Rancourt *et al.*² found coefficients of friction in the range 0.3 and 1.3 for the interface between metals and tibial cancellous bone. The results from the current study for stainless steel against bone are in this range as are the values from the study of Dammak *et al.*³ Davim and Marques⁴ found the average coefficient of friction to be lower than the values for the current study with 0.25 ± 0.02 for cancellous bone against steel with a surface roughness of $0.14 \pm 0.02 \mu\text{m}$. This surface roughness was similar to the surface grind ($0.13 \pm 0.04 \mu\text{m}$) used in this study, but the coefficient of friction was 0.595 ± 0.015 . The difference may be due to the different lubricants used, the roughness of the bone and the type of bone (cancellous or cortical).

This study has for the first time investigated the friction at the interface between bone and a surgical instrument. A detailed understand of these friction values are required for the design of new surgical instruments and the appropriate selection of materials and surface finishes. New surgical instruments will enable surgeons to have the appropriate control of the instruments. Based on the results of this study it is recommended that Silastic Q7-4780 silicone elastomer could be a potentially useful material for the design of bases of orthopaedic instruments that interface with bone.

Conclusions

A study to calculate the coefficient of static friction between bone and stainless steel or silicone elastomer, in the presence of a blood analogue solution, was conducted to assess the coefficient of friction provided by materials on a slippery bone surface. The main findings of this study are listed below.

1. Textured stainless steel surfaces generally have a higher coefficient of friction than silicone elastomers.
2. A stainless steel surface prepared by spark erosion, with a surface roughness of $8.9 \pm 1.6 \mu\text{m}$ exhibited higher value of coefficient of static friction than those textured by surface grinding (surface roughness $0.1 \pm 0.0 \mu\text{m}$) or sand blasting (surface roughness $2.2 \pm 0.1 \mu\text{m}$).
3. The coefficient of static friction of Silastic Q7-4780 silicone elastomer was not significantly different to those of the stainless steel samples obtained by surface grinding or sand blasting.
4. These findings will help in the design of new surgical instruments that interface with bone.

Conflict of interest statement

The authors had no conflicts of interest.

Funding

Funding for this project was provided by Sportsmed (UK) Ltd. Funding for the testing machine was provided by the Arthritis Research Campaign (now Arthritis Research UK).

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Tables

Table 1. Mean and standard deviation surface roughness (R_a) measurements.

Material	R_a (μm)
Silastic Q7 - 4720	1.08 ± 0.30
Silastic Q7 - 4750	1.45 ± 0.08
Silastic Q7 - 4780	1.73 ± 0.28
Surface Grind	0.13 ± 0.04
Sand Blast	2.19 ± 0.14
Spark Erode	8.94 ± 1.56

Table 2. Values of the coefficient of static friction, μ , mean (\pm standard deviation) for different interfaces obtained by fitting a regression line to the experimental data.

	Bone A	Bone B	Bone C	μ
Silastic Q7 – 4720	0.254	0.240	0.261	0.252 ± 0.011
Silastic Q7 – 4750	0.257	0.266	0.279	0.267 ± 0.011
Silastic Q7 - 4780	0.537	0.442	0.562	0.514 ± 0.063
Stainless steel - surface grind	0.599	0.578	0.607	0.595 ± 0.015
Stainless steel - sand blast	0.654	0.620	0.684	0.652 ± 0.031
Stainless steel - spark erode	0.785	0.713	0.728	0.742 ± 0.038

Figures

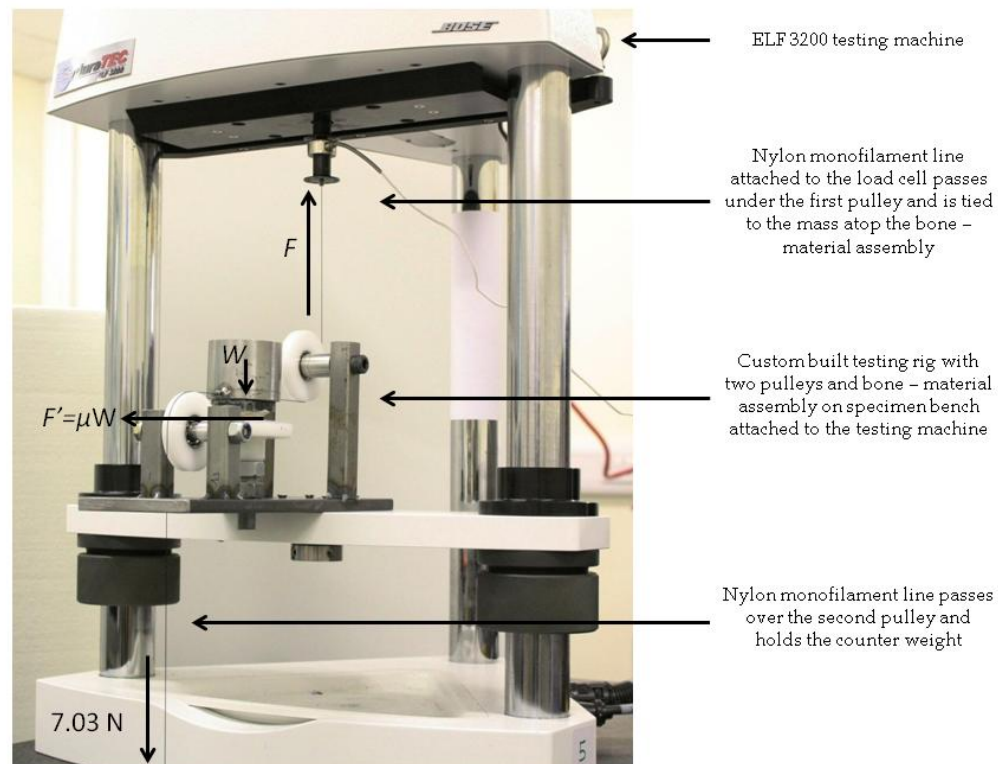


Figure 1: Experimental equipment to measure the coefficient of static friction. F is the force in the nylon monofilament line; F' is the breakaway force; μ is the coefficient of static friction; W is the compressive force.

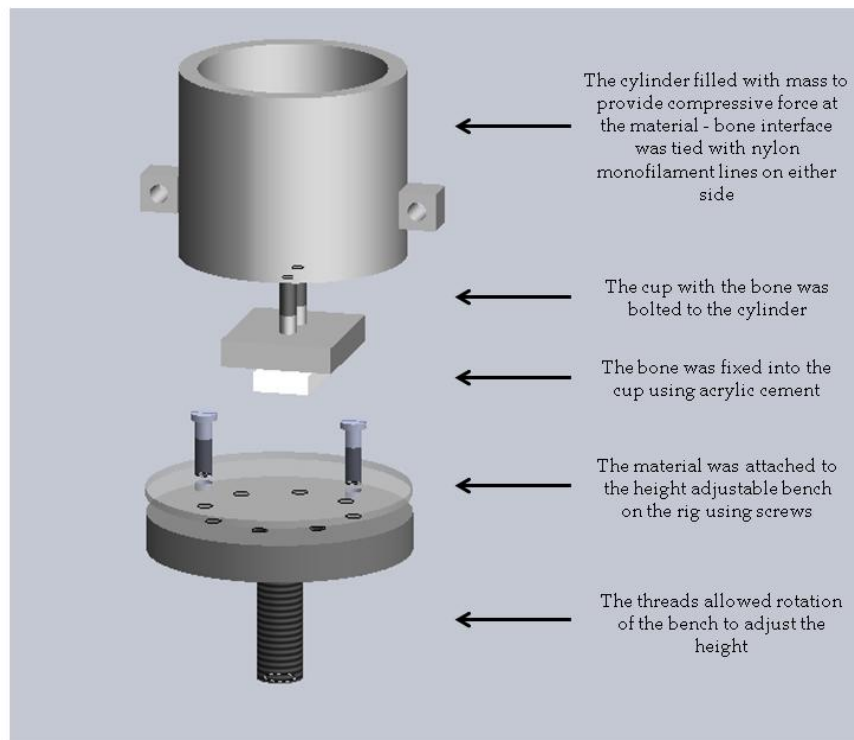


Figure 2: Exploded view of the assembly on top of the material.

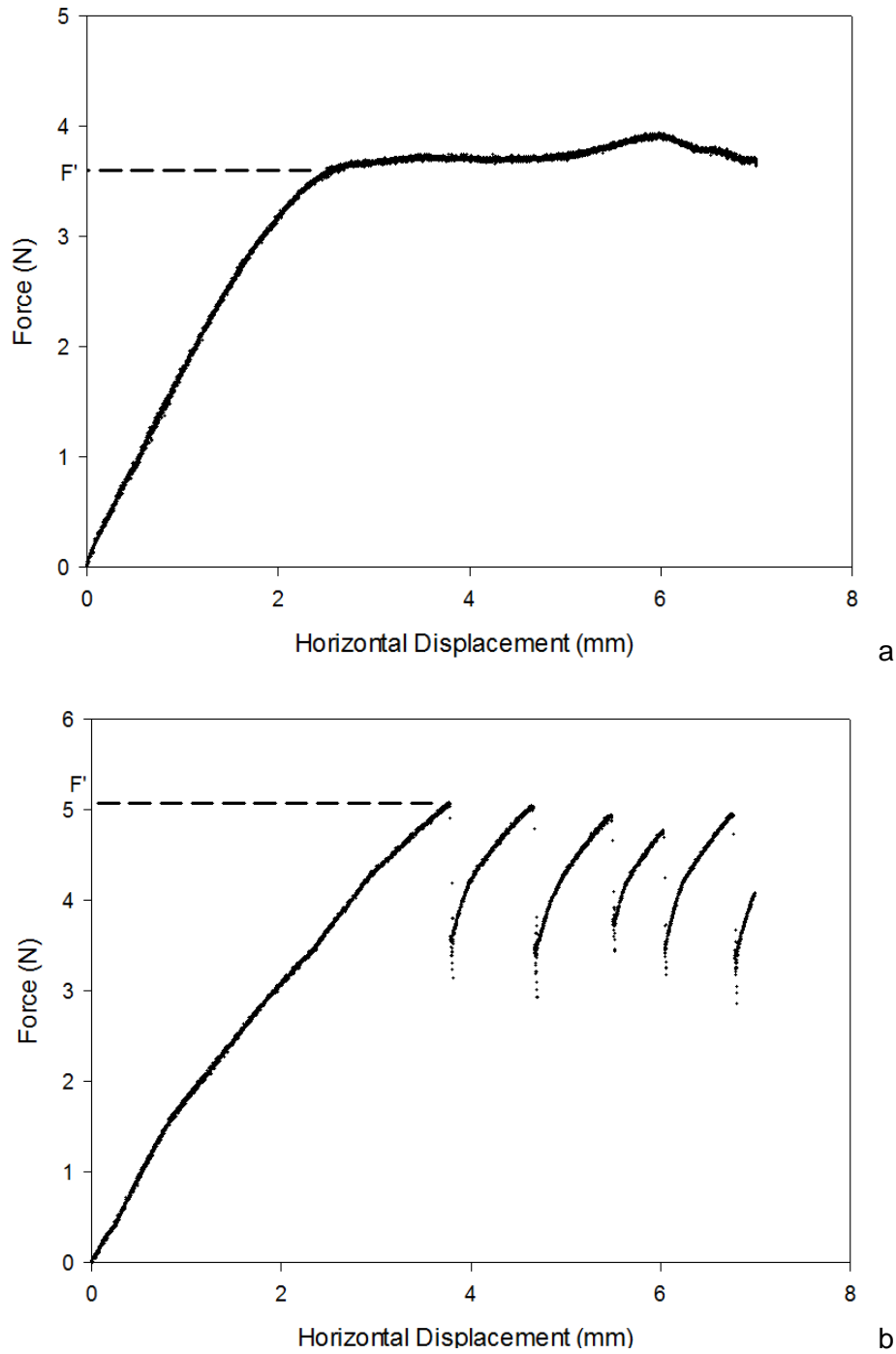
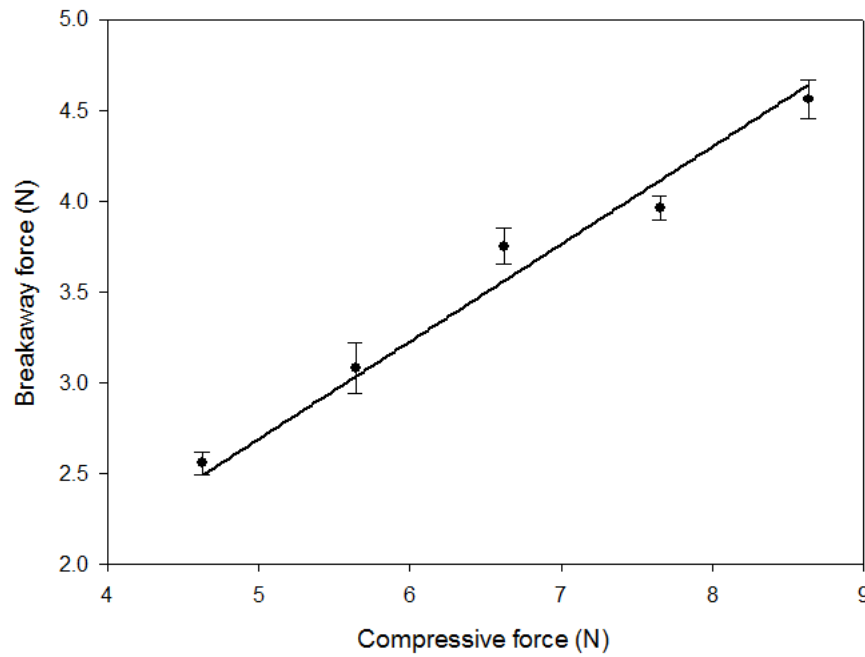
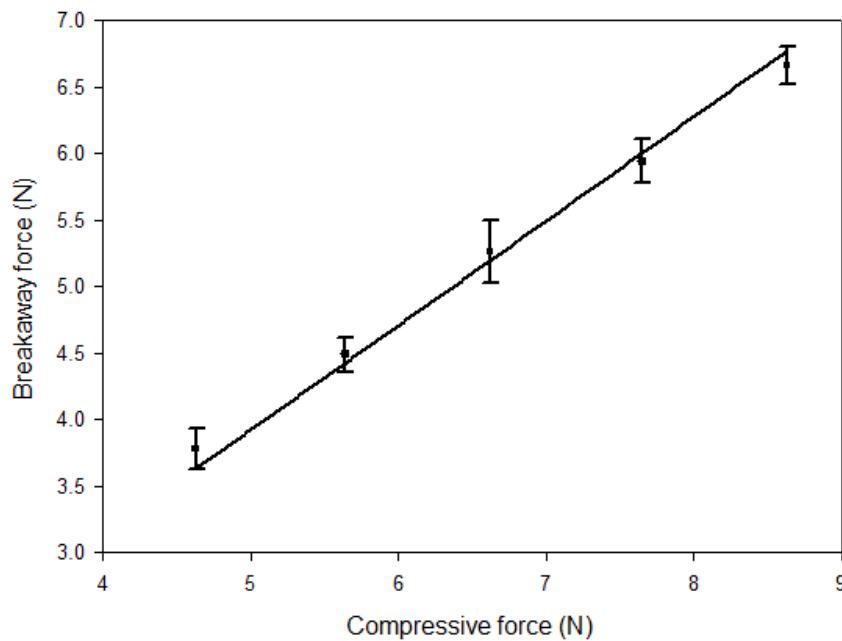


Figure 3. Horizontal displacement of bone specimen A for (a) Silastic Q7-4780 and (b) spark eroded stainless steel with a compressive load of 4.63 N. In (b) the interface exhibits stick-slip behaviour. In (a) the breakaway force, F' , was indicated by the first maximum in the curve; in (b) F' was the highest force before slippage occurred.



a



b

Figure 4. Values of Breakaway force, F' , plotted against compressive force for (a) Silastic Q7 – 4780 and (b) spark eroded stainless steel; in (b) the material combination had exhibited stick-slip behaviour. In both cases there is a significant linear regression; (a) $R^2 = 0.97$, $p < 0.0001$ (b) $R^2 = 0.99$, $p < 0.0001$. In both cases, the coefficient of static friction was taken to be the slope of the regression line.